NEW METHOD FOR MEASURE REGIONAL PULSE WAVE VELOCITY BY MEANS OF RF ULTRASONIC SIGNALS

Zbigniew TRAWIŃSKI

Department of Ultrasound, Institute of Fundamental Technological Research, Polish Academy of Sciences, 5B Adolfa Pawińskiego St., 02-106 Warsaw, POLAND

ztraw@ippt.gov.pl

The aim of this paper is to describe a non-invasive method of examination of the local pulse wave velocity. The measurements were carried out in the elastic silicon model of the artery immersed in water tank. Two synchronized ultrasonic apparatus VED with the ultrasonic radio frequency echoes acquisition for evaluation of the arterial elasticity, developed by the author, were used. The zero-crossing method was used for evaluating the pulse wave by measurements of the diameter changes of the model of the artery. The transit time between the waveforms of instant artery diameter at two measurement points, 5cm along the model of the artery was measured. The transit time was determined using the criteria of similarity of the first derivatives of the raising slopes of curves describing the instant vessel’s diameter changes in two measurement points of the model of the artery. The pulse wave velocity obtained by proposed two-point method was referred to the one-point method based of the modified Bramwell-Hill relation.

Keywords: ultrasound, local pulse wave velocity, model of artery

1. INTRODUCTION

The most frequently used methods of measuring the elasticity of vessels are the methods based on measurement of the change in the diameter of the artery as a result of blood pressure changes [1-9]. The functional dependence between the artery cross-section \( S \) and the pressure \( P \) inside the artery can be defined as follows [2, 4]:

\[
S = \frac{1}{g} \ln \left( \frac{P}{P_0} \right) \text{ [m}^2\text{]} \tag{1}
\]

where: \( g \) – constant coefficient, \( P_0 \) – reference pressure \((P>P_0>0)\).
Assuming, as the starting point, the dependence (1) we can determine coefficients \( g \) and \( P_o \) on the basis of the measuring the cross-section \( S \) for two characteristic points – values of artery cross-sections \( S_s \) and \( S_d \) which correspond to respective values of the blood pressure \( P_s \) and \( P_d \) for the systolic and diastolic phases of the heart. After inserting the values \( g \) and \( P_o \), obtained in the aforementioned manner, the formula (1) can be presented as:

\[
S(P) = S_d[1 + \frac{S_s - S_d}{S_d \ln(P_s / P_d)} \ln\left(\frac{P}{P_d}\right)] \quad [m^2].
\]

(2)

Assuming that the arterial cross-section is the circular shape, the \( S(P) \) can be replaced by the dependence \( \pi D_s^2(P)/4 \), where \( D \) is a diameter of an artery and dependence (2) can be transformed to:

\[
D_s^2(P) = D_o^2\left[1 + \frac{1}{\alpha} \ln\left(\frac{P}{P_d}\right)\right] \quad [m^2],
\]

(3)

where:

\[
\alpha = \frac{D_d^2}{D_s^2 - D_d^2} \ln\left(\frac{P_s}{P_d}\right),
\]

(4)

\( D_s \) and \( D_d \) are the artery diameter for the systolic \( P_s \) and diastolic \( P_d \) blood pressure respectively.

The coefficient \( \alpha \) was named the logarithmic coefficient of the artery wall stiffness [2].

For the examinations of coefficient \( \alpha \) the ultrasonic apparatus VED, developed by author, was used. The VED consists of the pulse wall tracking system which follows the movements of the artery walls with the accuracy of \( 7 \times 10^{-6} \)m. The examination were carried out on many healthy people of different ages [4, 5] and patients suffering from atherosclerosis [4]. The usefulness of the classification the examined people for the atherosclerosis high-risk group was presented in the author’s doctoral thesis [10].

The pulse wave is a phenomenon of the propagation of the blood pressure disturbances from the heart to the peripheral arteries in the body. The pulse wave velocity (PWV) depends of a blood density and the diameter of an artery. The pulse wave velocity is a very important indicator of the elasticity of the artery. Growing the value of the pulse wave velocity provides decreasing of the elasticity of the wall of the artery.

This paper presents a possibility of calculation the value of the coefficient \( \alpha \) by means of the value of the PWV. The PWV can be measured using many different methods. One of that was proposed in 2004 by Meinders et all. [3] where the pulse wave in the artery was determined by of the ultrasonic echoes RF signals reflected from arterial walls. In this article instead the correlation analysis the zero-crossing method has been proposed.
2. METHODOLOGY

The most well-known dependence describing the PWV is the formula introduced by Korteweg and Moens [3]:

$$PWV = \sqrt{\frac{Eh}{2\rho R}} \ [\text{m/s}],$$

(5)

where: $\rho$ - density of blood, $E$ – Young’s modulus of an artery wall, $h$ – vessel wall thickness, $R$ – internal artery radius when blood pressure is minimal.

In 1922 Bramwell and Hill [1] proposed the one-point method of determining the PWV based on the formula presented below:

$$PWV = \sqrt{\frac{S_d (P_s - P_d)}{\rho (S_s - S_d)}} \ [\text{m/s}].$$

(6)

Assuming that the artery cross-section has the shape of a circle, the dependence (6) can be expressed by:

$$PWV = \sqrt{\frac{D_d^2 (P_s - P_d)}{\rho (D_s^2 - D_d^2)}} \ [\text{m/s}].$$

(7)

Hence determining that:

$$\frac{D_d^2}{D_s^2 - D_d^2} = \frac{PWV^2 \rho}{P_s - P_d}$$

(8)

and inserting (8) into the formula (4), there is the coefficient $\alpha$ can be calculated:

$$\alpha = \frac{PWV^2 \rho}{P_s - P_d} \ln \left(\frac{P_s}{P_d}\right).$$

(9)

Therefore, the coefficient $\alpha$ can be calculated on the basis of measured the PWV, the density of blood $\rho$ and the systolic $P_s$ and diastolic $P_d$ blood pressure.

The new method for the measurement of the local PWV $c$ is presented in this work. For the determined the pulse wave, the instant diameter of the elastic model of the artery is measured. The two-point method for evaluation of the $c$ is based on the measurement of the transit time ($TT$) of the pulse wave travelled over the distance $L=5$cm in the model of the artery. In the medical practice, the measurement of the $c$ can be disturbed by the reflections from the arterial tree. To avoid these drawbacks, the two referred points at the beginning part of rising slope of the pulse wave, where no influence of the reflected pulse wave has been chosen. The method applied for the measurement of the $TT$, described in the author’s previous work [5], is based on the the criteria of similarity of the first derivatives of the rising slopes of the pulse wave. Therefore, the $c$ can be determined with the formula:
Different methods of measurement of the local \(PWV\) described by the author in the previous papers [5-9] were based on the ultrasonic measurement of the blood velocity with the use of the two-point Doppler method, as well as the one-point methods.

Another very important parameter of arterial condition is the elasticity modulus. The newest trends for ultrasonography applications concern the arterial elastography. The most of them present only the strain of the vessel walls, but more indicative parameter from diagnostic point of view is the Young’s modulus \(E\) of the artery wall. There is very difficult to measure the Young’s modulus of a human artery in the non-invasive way in the clinical conditions. The main problem concerns a non-invasive measurement of a thickness \(h\) of an arterial wall.

For the demonstrate of the possibility of presented methods and equipments for elastography examinations of arterial walls, there was demonstrate of the Young’s modulus calculation using two approaches. First, according to the formula (5), for calculating the Young’s modulus \(E\) it was required to knowing the values of: \(\rho\) - density of blood, \(h\) – vessel wall thickness and \(R\) – internal artery radius when blood pressure is minimal. At second, obtaining directly from strain-stress test. The application of the VED apparatus permits for measurement both the \(PWV\), \(R\) and \(h\) in the model of the artery.

3. EXPERIMENTAL SETUP

The measuring setup consisted of the bottom and the overflow tank, both made of polyethylene and the rotor water pump. The level of distilled water in the overflow tank was constant. Water from the overflow tank flowed due to gravity through the silicone pipe (diameter=13mm) into the computer-controlled piston pump and through artificial cardiac valves to the silicon arterial model placed in the aquarium filled with distilled water. The level of water in the aquarium was 3cm above the level of the model. The rhythm of the pump was simulated the conditions in the left ventricle of the human heart (pulse frequency equaled 1Hz). The diagram of the experimental setup for the measurement of the local \(PWV\) in the elastic model of the artery by means of the two-point method is presented in Fig. 1.

The measurements of local \(PWV\) were performed using measuring system based on two ultrasonic VED apparatus. The frequency of transmitted ultrasound was equal to 6.75MHz and 9kHz for the repetition frequency. The acquisition of the ultrasonic RF signal was performed with 14 bit precision and with the sampling frequency equal to 62.5MHz, simultaneously in the two channels, by means of the Signatec PC card.
Fig. 1. Diagram of experimental setup: R1 – upper overflow tank, R2 – bottom tank, Piston pump – SPA3891 Vivitro Co., P – rotor pump, V – artificial cardiac valve, C – windkessel model, Rp – hydraulic resistor, Elastic model – silicone model of the artery, VED1 and VED2 ultrasonic apparatus, 8 – aquarium, 10 - ultrasonic measuring heads, PM – pressure gauge with catheter, S – computer driver for the piston pump, RF PDA-14 – two channel RF Signatec PC card for the acquisition of RF signals, PC – IBM-PC.

The internal diameter of silicon model of the artery was equal 7.5mm, the wall thickness of used pipe was equal to 1.25mm and the length was equal to 96cm (after initial stretching up to 100cm).

4. RESULTS

Fig. 2 presents normalized traces of the pulse waves represented by the diameter changes of the elastic model of the artery at the two points located on borders of the 5cm segment of the pipe.
Fig. 2. Normalized pulse waves (the internal diameter D1 and D2) at the two measuring points of the model of the artery, occurring as a result of the forced cyclical change of the liquid pressure inside the vessel.

In the same conditions, taking into account values of the $D_s$ and $D_d$ and diastolic $P_s$ and diastolic $P_d$, the PWV was determined by means of the one-point method, in accordance with Bramwell-Hill dependence (6). The measurement results of the $c$ and PWV, averaged over 30 measurements, are presented in Table1.

Tab. 1. PWV determined by one-point and two-point method.

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<tr>
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<th>One-point method</th>
<th>Two-point method</th>
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<tr>
<td>PWV [m/s]</td>
<td>$28.1 \pm 0.12$</td>
<td>$32.4 \pm 0.13$</td>
</tr>
<tr>
<td>$c$ [m/s]</td>
<td></td>
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The Young’s modulus of the material from which the examined elastic model of the artery was made, calculated using the formula (5) is equal to $E = 5.03\text{MPa}$.

The value of the Young’s modulus becomes from the stress–strain relation determined experimentally at the deformation up to 15% is equal to $E = 4.97\text{MPa}$. It gives the relative Young’s modulus measurement error at the level of 1.2%.

5. DISCUSSION AND CONCLUSIONS

This paper presented results of the examination of the elasticity of the carotid artery walls for the classification people for the high group risk for atherosclerosis.

The main conclusion of the authors examination of the coefficient $\alpha$ is that decreasing of the arterial elasticity is one of the primary symptoms of early state of the atherosclerosis.

Also the new technique of determined of the logarithmic coefficient of the artery wall stiffness $\alpha$, on the base of the value of the $PWV$ was presented.

The main advantage of proposed the two-point method for arterial local pulse wave velocity measurement are the avoidance of the blood pressure measurement and the errors arising from this measurement, what usually contribute a large additional error at a level of $\pm 10\%$. 
The application of the zero-crossing method for determining the pulse wave propagated in the model of the artery, in comparison with the correlation method, enables to avoid the errors arising from the ambiguity of determining constant integrals, which can cause ambiguous trends (a bias).

Using the criteria of the similarity of the first derivatives of rising slopes of the pulse waves in two points located near each other, there is the best choice of the reference points for calculating the \( \text{PWV} \) because it is the best fitted regions of the pulse waves, where reflected pulse waves not exist.

Both for the \( c \) and the \( \text{PWV} \) the coefficient of variation (standard deviation/mean) were 0.4% it means that measurement repeatability was very good.

The values of the local pulse wave velocity \( c \) and the \( \text{PWV} \) differed by 13%. It could be considered an acceptable error in comparison with the results of the clinical research published by various authors.

The method of examination of the local pulse velocity presented in this paper requires further verification on a large group of persons.

REFERENCES